EFFECTS OF AGING ON PATELLOFEMORAL JOINT STRESS DURING STAIR NEGOTIATION ON CHALLENGING SURFACES

by

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DEDICATION

I would like to dedicate this thesis to my family, and in particular my mother Jodie Hunt, for the continuous support in my personal, academic, and professional endeavors. I would also like to dedicate this moment to Tina Freeman for her omnipresent guidance throughout my academic career and providing the means necessary to accomplish this feat.

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ABSTRACT

Introduction: Patellofemoral pain is an incessant lower limb musculoskeletal disorder that may be underreported in older adults. During common locomotor activities, such as when negotiating stairs, older adults (over the age of 65 years) adopt knee biomechanics reported to increase patellofemoral pain. Negotiating stairs with a challenging surface, such as uneven or slick, may place greater demand on the knee and further exacerbate joint biomechanics related to PFJ stress. Yet, it is unknown if older adults exhibit increases in patellofemoral joint (PFJ) stress when negotiating stairs with challenging surfaces. **Purpose:** The purpose of this study was to examine the effect of age (young and older adults) and surface (normal, slick, and uneven) on the magnitude and temporal waveform of patellofemoral joint stress during stair ascent and descent tasks. **Methods:** Two cohorts (12 young: ages 18-25 years; 12 older: over 65 years) had knee biomechanics quantified after they ascended and descended 18.5 cm stairs on normal, slick, and uneven surfaces at a self-selected speed. **Statistical Analysis:** Peak of stance (0-100%) PFJ stress and associated components (including PFJ reaction force and contact area, and knee flexion angle and moment) were submitted to a two-way RM ANOVA to test the main effects of and interaction between age (young vs old) and surface (normal, slick, and uneven). A statistical parametric mapping two-way ANOVA was used to determine main effects of and interaction between age and surface for the PFJ stress waveform. **Results:** During the stair ascent, older adults exhibited greater PFJ stress from 56 to 84% of stance ($p < 0.001$), which may be attributed to the greater PFJ stress-time integral ($p = 0.004$) and later peak PFJ stress ($p = 0.024$) compared to young adults. Additionally, a significant age by surface interaction was observed for time of peak PFJ stress ($p = 0.041$) during stair ascent, where older adults exhibited a later peak PFJ stress compared to young adults ($p = 0.008$), and later peak PFJ stress compared to normal and slick surface (both: $p = 0.014$). Surface impacted PFJ stress waveform (all: $p < 0.001$), but not magnitude ($p > 0.05$) during both stair ascent and descent. During stair ascent on the uneven surface, participants exhibited smaller PFJ stress from 8 to 25% of stance compared to normal surface, but greater PFJ stress from 57 to 90% and 49 to 77% of stance compared to the normal and slick surfaces (all: $p < 0.001$). On the uneven surface, participants exhibited a greater PFJ stress-time integral (both: $p = 0.010$) compared to the normal and slick surfaces. During stair descent, on the uneven surface, participants only exhibited greater PFJ stress-time integral ($p = 0.017$) compared to slick surface, while PFJ stress was smaller from 5 to 18% of stance, but greater stress from 92 to 99% of stance (both: $p < 0.001$) on the slick compared to the normal surface.

Conclusion: Older adults are more likely to exhibit knee biomechanics related to PFJ pain development when navigating stairs. Specifically, the larger, later PFJ stress exhibited by older adults when ascending, but not descending the stairs may increase loading of the joint's articular cartilage and increase risk of developing PFJ pain. Yet, all participants exhibited alterations in knee biomechanics that may lead to greater PFJ stress when negotiating stairs with slick and uneven surfaces.

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PFP	Patellofemoral Pain
PFJ	Patellofemoral Joint
vGRF	Vertical Ground Reaction Force
PFJRF	Patellofemoral Joint Reaction Force
PFJCA	Patellofemoral Joint Contact Area
ADLs	Activities of Daily Living
GRF	Ground Reaction Force
3D	Three-dimensional
SPM	Statistical Parametric Mapping
PS	Peak of Stance

CHAPTER ONE: INTRODUCTION

Musculoskeletal pain is a common, costly problem for older Americans. Treating the nearly 60% of older adults (over the age of 65 years) affected by musculoskeletal pain costs $635 billion annually, but still poses a significant health risk for the inflicted.[1,2] Musculoskeletal pain and associated disorders are reportedly associated with falls, frailty, reduced mobility, and impaired cognitive function in older adults.[3-8] Yet, musculoskeletal pain may be undertreated and underreported in older adults, as there is commonly a false belief that pain is associated with normal aging and/or individuals have an inability to communicate pain.[9-12] Patellofemoral joint pain (PFP), which is an incessant lower limb musculoskeletal disorder that affects nearly 23% of adults, may be routinely undertreated and/or underreported in older adults.[13] Although elevated incidence of PFP is evident starting at 30-49 years and continually increases until about 60 years of age[14], with higher rates among women and physically active individuals[13], the prevalence of PFP in older adults remains relatively unknown. Identifying and treating knee pain, specifically PFP, may improve older adult mobility and prevent the loss of independence and impaired quality of life commonly associated with aging.[15-17] However, it is unknown whether aging leads to alterations in lower limb biomechanics, particularly at the knee, that may increase the likelihood of PFP.

Patellofemoral joint pain is an overuse musculoskeletal disorder that stems from multiple etiologic factors, including alterations of lower limb biomechanics during locomotion. The development of PFP results from repeated application of elevated forces on the articular cartilage of the patellofemoral joint (PFJ) leading to degeneration of the soft-tissue and subsequent pain. Stress, or force per unit area, of the PFJ is traditionally reported as intensity (i.e., peak magnitude) or stress vs time profiles (area under the stress-time curve or stress-time integral), and may provide key insight into tissue damage that leads to PFP.[18,19] Yet, previous experimental evidence exploring differences in PFJ stress between PFP and healthy populations is inconclusive. Individuals with PFP tend to walk slower, and exhibit a smaller peak vertical ground reaction force (vGRF) as well as peak knee extension moment, biomechanical changes reported to decrease peak patellofemoral joint reaction force (PFJRF) and PFJ stress; however, these individuals also exhibit concurrent reductions in patellofemoral joint contract area (PFJCA) that may elevate PFJ stress.[20,21] The cautious gait adopted by PFP individuals produces larger, insignificant increases in peak PFJ stress, but large, significant increases of PFJ stress-time values (greater than 200%) compared to healthy controls.[20,23] However, when walk velocity in controlled, individuals with PFP exhibit significantly greater peak and stress-time PFJ stress compared to healthy controls, which may be attributed to increases in their peak knee extension moment, particularly in late stance or swing.[20] Brechter and Powers (2002) observed greater knee extension moment for PFP individuals during terminal stance during walking[20], and thus, traditional PFJ stress (i.e., discrete peak and stress-time integral) measures may fail to identify biomechanical differences that lead to increases in PFJ stress and PFP. Considering PFP individuals and older adults, are observed to exhibit temporal or waveform difference (i.e., time of peak and local peaks) in knee biomechanics related to PFJ stress, waveform analysis of knee biomechanics related to PFP may be necessary to differentiate age-related changes in PFJ loading.

Older adults reportedly adopt cautious gait strategies during locomotive activities of daily living (ADLs), such as walking and stair negotiation. Older adults cautious gait strategies during ADLs, include walking slower with a flexed knee and greater quadriceps-hamstring co-contraction as well as decreases in peak knee extension moment.[24-27] The cautious gait strategies exhibited by older adults may also be necessary to compensate for progressive reductions in strength and joint stability associated with aging.[28-31] Although these biomechanical adaptations theoretically reduce PFJRF and subsequently PFJ stress, cartilage degeneration associated with normal aging may also reduce PFJCA leading to substantial increases in PFJ stress for older adults.[32-34] Further, during walking, older adults exhibit increases in quadriceps contraction and knee extension moment in late stance similar to PFP individuals, which may lead to concurrent increases in PFJ stress-time profiles.[35-36] Yet, it is currently unknown whether older adults exhibit greater magnitudes or temporal differences in PFJ compared to their younger counterparts.

Negotiating stairs (both ascending and descending) is more physically demanding than level walking. Specifically, at the knee, stair ascent and descent requires 50% more flexion range of motion and 50% greater peak knee extension moment compared to level walking.[37-41] This increased demand results in two and four times greater PFJ stress, which may be attributable to large increases in peak PFJFRF and knee joint moments necessary to safely negotiate stairs.[22,42,43] Age-related changes (i.e., reductions) in muscle strength presented by older adults may result in maladaptive increases in lower limb joint moments in general, but knee moments specifically, to safely negotiate stairs.[40,44] These changes may lead to further increases in PFJ stress for older adults, however, to date, the effect of age on PFJ stress during stair ascent and descent remains relatively unknown.

Challenging environmental conditions, such as slick or uneven surfaces during ADLs, may further alter older adults' gait and increase PFJ stress. When navigating a slick or uneven surface, individuals, particularly older adults, tend to walk slower with shorter, more variable strides, and increase muscle activation to provide the stability necessary to prevent a fall and protect a joint from injury.[45-52] When older adults navigate a challenging surface, they reportedly exhibit greater changes in knee flexion than their younger counterparts, and may further increase PFJ stress.[53] Yet, the effect of slick and uneven surfaces on older adult lower limb biomechanics, particularly PFJ stress, remains largely unknown. With that in mind, this study will seek to investigate the effect of surface (slick and uneven) and age (young and older adults) on PFJ stress during stair negotiation (both ascend and descend).

Specific Aims

<u>Specific Aim 1</u>

To examine patellofemoral joint stress for young and older adults during stair negotiation. Specifically, this study will quantify magnitude and temporal (i.e., waveform) differences in stance phase patellofemoral joint stress, and its associated components (including patellofemoral reaction force and contact area as well as knee flexion angle and extension moment) for young (between 18 and 25 years) and older adults (over 65 years) ascending and descending 18.5 cm stairs at a self-selected speed.

Hypothesis 1.1

Older adults will not exhibit a significant difference in the magnitude of patellofemoral joint stress and associated components compared to young adults during the stair ascent and descent.

Hypothesis 1.2

During the stair ascent and descent, older adults will exhibit significant waveform differences in patellofemoral joint stress and associated components compared to young adults.

Significance

Understanding magnitude and waveform changes of patellofemoral joint stress with age will aid in the reduction of musculoskeletal pain, particularly patellofemoral pain, for older adults. Determining the specific maladaptive knee biomechanics adopted by older adults during stair negotiation that increase the risk for patellofemoral joint pain will provide clinicians knowledge of explicit biomechanical parameters to target for beneficial reductions in knee pain.

Specific Aim 2

To examine patellofemoral joint stress for young and older adults when they negotiate stairs with challenging surfaces. Specifically, this study will quantify magnitude and waveform differences in stance phase patellofemoral joint stress, and its associated components (including patellofemoral reaction force and contact area as well as knee flexion angle and extension moment) when young (between 18 and 25 years) and older adults (over 65 years) ascend and descend 18.5 cm stairs on normal, slick, and uneven surfaces at a self-selected speed.

Hypothesis 2.1

During stair ascent and descent, all participants will exhibit a significant increase in the magnitude of patellofemoral joint stress and associated components on slick and uneven compared to normal surface, but significant differences between older and young adults will not be observed.

Hypothesis 2.2

There will not be significant waveform differences in patellofemoral joint stress and associated components between each surface, but older adults will exhibit significant waveform differences on each surface compared to young adults.

Significance

Determining whether challenging surfaces, such as slick and/or uneven stairs, impact patellofemoral joint stress will provide the knowledge necessary to decrease patellofemoral joint pain. In particular, it will provide clinicians critical insight into specific biomechanical strategies adopted on challenging stair surfaces to target for beneficial reductions in knee pain.

CHAPTER TWO: LITERATURE REVIEW

The following section aims to detail aging, specifically the 1) aging population, 2) musculoskeletal pain related to aging, 3) patellofemoral pain and mechanics, and 4) lower limb biomechanics of older adults.

Aging

Older Adult Population

The older adult population has rapidly grown since the turn of the 20th century attributable to lower fertility and increased longevity. The older adult population in the United States grew from 3.1 million in 1900 to 35 million in 2000, a trend that continues today.[54] In 2016, the American Community Survey reported over 49 million individuals over the age of 65, accounting for approximately 13% of the total population.; however, by the year 2030, projections suggest older adults will exceed 72 million, representing nearly 19% of the total US population.[55] The implication of an aging nation provides also provides a significant financial burden on the healthcare system. Older adults average medical expenditures are more than 2.6 times the national average, accounting for one-third of US medical spending.[56] Although over 65% of older adult health care costs are subsidized by the government and about 13% is covered through private insurance, the remaining 20% is financed out-of-pocket resulting in an average of over $5,700 per person in 2015, up almost 40% since 2005.[57] These out-of-pocket expenditures are 75% higher compared to the general population ($4,342) as older adults spend 13% of their total expenditures on health compared to 8% of all consumers.[57] Understanding the cost and implications that occur with an aging population provides context to explore changes that occur with normal aging in general, or the musculoskeletal system in general.

Musculoskeletal Pain

Musculoskeletal disorders provide a significant physical and financial burden on the general population. More than one out of every two individuals age 18 and over in the United States are affected by musculoskeletal disorders resulting in costs estimated at $980 billion per year in 2014, with this burden increasing annually.[58] Musculoskeletal pain and associated disorders cause significant risk to the maintenance of health in older age as they are associated with falls, frailty, reduced mobility, and impaired cognitive function.[59-64] Musculoskeletal pain affects up to 60% of people aged 65 and older costing up to $635 billion annually.[1,2] Despite the implications, musculoskeletal pain for older adults may be undertreated and underreported due to various psychosocial factors. These factors that may affect pain reporting include, but are not limited to false belief that pain is associated with normal aging, lack of identification of pain, cognitive impairment and inability to communicate pain, and potential fear or embarrassment about pain.[64-67]

Knee pain is highly prevalent affecting approximately 25% of adults accounting for nearly 4 million healthcare visits annually.[68,69] The prevalence of knee pain has increased almost 65% over the past 20 years and increases universally with age. Knee pain in older adults is associated with reduced strength, balance, and physical function resulting in significant reductions in mobility, independence, and quality of life.[15,70-73] Nearly 50% of older adults report knee pain annually, with at least 50% of those reporting some restriction of activities of daily living.[74,75] Pain at the patellofemoral joint has been reported to represent up to 33% of all knee related injuries for individuals between the ages of 10 and 60.[14]

Musculoskeletal Disease

Patellofemoral Pain

Patellofemoral pain (PFP) is anterior knee pain characterized by increases in compressive force on the patellofemoral joint (PFJ). This condition affects nearly 23% of adults with higher incidence in women compared to men, and specifically, physically active individuals and military personnel.[13] Previous reports suggest that PFP is most prevalent in individuals between the ages of 16-25 and individuals below the age of 35 were at greater risk of developing PFP compared to older adults.[76,77] However, epidemiological trends show a linear increase in PFP incidence from 20 years of age to 60 years of age, but the prevalence of PFP in older adults is unclear and may be underreported.[14] Patellofemoral pain results in articular cartilage degermation related to the magnitude, duration, and frequency of applied load, thus increased exposure to locomotive tasks due to aging predisposes older adults to cartilage degenerative disease and pain.[78-80] Further, PFP has been linked to the development of patellofemoral osteoarthritis, resulting in significant burden on healthcare systems and the individual alike.[81,82] Patellofemoral osteoarthritis has become increasingly common over the last 20 years, with the highest prevalence in individuals between the ages 50-70.[83]

Patellofemoral Pain Biomechanics

Despite the high prevalence, the specific pathomechanics of PFP remains unclear. Traditionally, abnormal patellar alignment and/or tracking were thought to be the primary cause; however, patellar malalignment is only present in a subset of individuals with PFP as many individuals with patellar abnormalities never develop PFP symptoms.[84] Recent literature proposes the "theory of tissue homeostasis" as a model for the development of PFP. Dye (2005) suggests any alterations in tissue homeostasis that exceeds the load acceptance capacity of the PFJ (i.e., structural abnormalities or repetitive overloading) results in symptomatic bone and soft tissue damage, and subsequent pain.[85] The etiology of PFP is considered multifactorial with potential causes including overuse, overload, muscular dysfunction, and abnormal lower extremity biomechanics during gait. Specifically, local joint factors include patellar maltracking, quadriceps weakness, delayed vastus medialis activation, and soft tissue inflexibility (i.e., quadriceps, gastrocnemius, iliotibial band, and hamstrings).[84] Gait aberrations include excessive hip adduction and internal rotation, femur internal rotations, and foot pronation as well as increased vertical ground reaction force and decreased knee flexion angle at initial contact.[86-90] Other factors that may contribute to overloading the joint capacity are increases in activity duration, frequency, or intensity as well as irregular surfaces.[91] Although these factors may not result in immediate damage, repetitive elevated stresses may result in tissue damage over time and subsequent pain at the joint.[85] Thus, activities that result in increased PFJ stress (i.e., squatting, running, or negotiating stairs) or challenging surfaces (i.e., slick or uneven) increases the risk and symptoms of PFP.

Effects of Aging

Physiological

Musculoskeletal changes that occur with normal aging, such as reduction of muscle strength and increase in joint stiffness, has major implications on physical function in older adults during activities of daily living (ADLs). In general, older adults operate up to 22% closer to their relative lower limb muscle strength despite adopting a slower walking speed, reflecting a higher relative cost of mobility that may not be sustainable for long periods of locomotion.[92,93] Further, there is a U-shaped relationship between speed and energy cost in which healthy older adults produce an upward shift resulting in a 15-25% increase in energy cost compared to young adults at any walking speed.[94-96] This increase in energy cost may be the result of greater dynamic instability during gait, lower limb joint mechanical work, and muscle co-activation.[95,97-99] The biomechanical changes associated with increased energy expenditure may increase knee biomechanics related to PFJ stress and increase risk of PFP in older adults.

Spatiotemporal Changes

Gait parameters may be used to assess physical function and quality of life in older adults and can be evaluate risk of neurological disorders, falls, and early mortality.[30,100-103] Older adults reportedly walk slower as a result of decreased stride length, increased stance time, and longer double-support phase compared to young adults.[24,104] Further, older adults exhibit large increases in gait variability, including cadence, stride length, and stride width, resulting in increased risk for frailty, falling, and neurodegenerative disease compared to their younger counterparts.[105] Changes to spatiotemporal parameters of gait in older adults result in a significant changes to force attenuation, limb loading, and joint kinematics.

Ground Reaction Forces

Ground reaction force (GRF) during gait can provide insights on limb loading and physical function in older adults. The relationship between gait speed and GRFs is well understood, as slower walking speeds produce lower GRFs for all populations. Older adults, who naturally adopt a slower gait to prevent injury and falls, report significantly smaller vertical GRFs, specifically at the first and second peak compared to young adults.[106,107] Further, older adults also exhibit significantly lower horizontal GRF during the propulsion phase, likely to maintain balance and prevent joint injury.[106] Muscle strength may also be used as a predictor for walk speed and GRF during gait, as low strength older adults walk even slower and exhibit significantly lower vertical GRF during the weight acceptance phase of gait compared to their stronger counterparts.[108] These age-related changes to GRFs predispose older adults to mobility limitation, disability, and loss of independence as well as reflects compensatory strategies that alter lower limb biomechanics and joint loading.[109-111]

Joint Redistribution

Due to reductions in muscle strength that occur with normal aging, older adults are reported to adopt a distal to proximal shift in the relative contribution of the lower extremity joints during gait. This compensatory gait strategy results in abnormal joint loading that increases the risk for musculoskeletal disorders. The proximal joint redistribution greatly increases the role of the hip during locomotion for older adults. When controlling for walk speed, the hip has a significantly greater range of motion, flexion at heel-strike, and peak flexion, but less hip extension compared to young adults during walking.[106] Further, older adults have been reported to significantly increase power generation at the hip compared to young adults during walking with increases in angular impulse and work as high as 58% and 279%, respectively.[112,113] At the knee, older adults tend to be more flexed and exhibit lower knee extension moments with up to 50% less angular impulse and 40% less work compared to young adults.[112,113] For the ankle, older adults are more plantarflexed with less range of motion while generating up to 23% and 30% less angular impulse and work, respectively, compared to young adults.[55,112,113] This phenomena is often described in walking, however, similar findings extend to stair negotiation comparing young and older adults.[114] Although this modified gait strategy is intended to preserve balance and prevent injury, it provides unique lower extremity biomechanics in general, and at the knee specifically, that may increase PFJ stress in older adults.

<u>Kinematics and Kinetics</u>

Older adults adopt compensatory gait strategies at the knee comparable to individuals with PFP. Specifically, older adults walk slower, increase flexion, increase co-contraction, and decrease peak knee extension moments compared to healthy young adults.[24-27] These biomechanical strategies should hypothetically reduce PFJ contact force and subsequent PFJ stress; however, cartilage degeneration associated with the normal process of aging should reduce PFJ contact area should result in counteractive increases in PFJ stress for older adults.[32-34] Older adults, and individuals with PFP alike, have been reported to increase knee extension moments in late stance which would result in a subsequent increase in PFJ stress during terminal stance and reflect a larger stress-time profile.[20,114] However, it is unknown whether older adults exhibit greater magnitude or temporal differences in PFJ stress compared to young adults.

CHAPTER THREE: MANUSCRIPT

Introduction

Musculoskeletal pain is a common, costly problem for older Americans. Treating the nearly 60% of older adults (over the age of 65 years) affected by musculoskeletal pain costs $635 billion annually, but still poses a significant health risk for the inflicted.[1,2] Musculoskeletal pain and associated disorders are reportedly associated with falls, frailty, reduced mobility, and impaired cognitive function in older adults.[3-8] Yet, musculoskeletal pain may be undertreated and underreported in older adults, as there is commonly a false belief that pain is associated with normal aging and/or individuals have an inability to communicate pain.[9-12] Patellofemoral joint (PFJ) pain, which is an incessant lower limb musculoskeletal disorder that affects nearly 23% of adults, may be routinely undertreated and/or underreported in older adults.[13] Although elevated incidence of PFJ pain is evident starting at 30-49 years and continually increases until about 60 years of age[14], with higher rates among women and physically active individuals[13], the prevalence of PFJ pain in older adults remains relatively unknown. Identifying and treating knee pain, specifically PFJ pain, may improve older adult mobility and prevent the loss of independence and impaired quality of life commonly associated with aging.[15-17] However, it is unknown whether aging leads to alterations in lower limb biomechanics, particularly at the knee, that may increase the likelihood of PFJ pain.

Patellofemoral joint pain is an overuse musculoskeletal disorder that stems from multiple etiologic factors, including alterations of lower limb biomechanics during locomotion. The development of PFJ pain results from repeated application of elevated forces on the articular cartilage of PFJ leading to degeneration of the soft-tissue and subsequent pain. Stress, or force per unit area, of the PFJ is traditionally reported as intensity (i.e., peak magnitude) or stress vs time profiles (area under the stress-time curve or stress-time integral), and may provide key insight into tissue damage that leads to PFJ pain.[18,19] Yet, previous experimental evidence exploring differences in PFJ stress between PFJ pain and healthy populations is inconclusive. Individuals with PFJ pain tend to walk slower, and exhibit smaller peak vertical ground reaction force (vGRF) and peak knee extension moment, or biomechanical changes reported to decrease peak PFJ reaction force and PFJ stress. However, these individuals also exhibit concurrent reductions in PFJ contact area that may elevate PFJ stress.[20,21] The cautious gait adopted by PFJ pain individuals produces larger, insignificant increases in peak PFJ stress, but large, significant increases of PFJ stress-time values (greater than 200%) compared to healthy controls.[20] However, when walk velocity is controlled, individuals with PFJ pain exhibit significantly greater peak and stress-time PFJ stress compared to healthy controls, which may be attributed to increases in their peak knee extension moment, particularly in late stance or swing.[20] Brechter and Powers (2002), for example, observed greater knee extension moment for PFJ pain individuals during terminal stance.[20] Thus, traditional PFJ stress (i.e., discrete peak and stress-time integral) measures may fail to identify biomechanical differences that lead to increases in stress and pain at the joint. Considering PFJ pain individuals and older adults, are observed to exhibit temporal or waveform difference (i.e., time of peak and local peaks) in knee biomechanics related to

PFJ stress, waveform analysis of knee biomechanics related to PFJ pain may be necessary to differentiate age-related changes in PFJ loading.

Older adults reportedly adopt cautious gait strategies during locomotive activities of daily living (ADLs), such as walking and stair negotiation. Specifically, during ADLs, older adults walk slower with a flexed knee and greater quadriceps-hamstring co-contraction, but smaller peak knee extension moment.[24-27] These cautious gait strategies exhibited by older adults may be necessary to compensate for reductions in lower limb strength and joint stability associated with aging.[28-31] Although these biomechanical adaptations theoretically reduce PFJ reaction force and subsequently PFJ stress, cartilage degeneration associated with normal aging may also reduce PFJ contact area leading to substantial increases in PFJ stress for older adults.[32-34] Further, during walking, older adults exhibit increases in quadriceps contraction and knee extension moment in late stance similar to PFJ pain individuals, which may lead to concurrent increases in PFJ stress-time profiles.[35-36] Yet, it is currently unknown whether older adults exhibit greater magnitudes or temporal differences in PFJ stress compared to their younger counterparts.

Negotiating stairs (both ascending and descending) is more physically demanding than level walking. Specifically, at the knee, stair ascent and descent requires 50% more flexion range of motion and 50% greater peak knee extension moment compared to level walking.[37-41] This increased demand results in two and four times greater PFJ stress and may stem from large increases in peak PFJ reaction force and knee joint moments necessary to safely negotiate stairs.[22,42,43] Age-related changes (i.e., reductions) in muscle strength exhibited by older adults may result in maladaptive increases in lower limb joint moments in general, but knee extension moment specifically, to safely negotiate stairs.[40,44] These changes may further increases in PFJ stress for older adults, however, the effect of age on PFJ stress during stair ascent and descent remains relatively unknown.

Challenging environmental conditions, such as slick or uneven surface during ADLs, may further alter older adults' gait and increase PFJ stress. When navigating a slick or uneven surface, individuals, particularly older adults, tend to walk slower with shorter, more variable strides, and increase muscle activation to provide the stability necessary to prevent a fall and protect a joint from injury.[45-52] Older adults also reportedly exhibit greater changes in knee flexion when navigating a challenging surface than their younger counterparts, which may further increase PFJ stress.[53] Yet, the effect of slick and uneven surfaces on older adult lower limb biomechanics, particularly PFJ stress, remains largely unknown. With that in mind, this study will seek to investigate the effect of age (young and older adults) and surface (slick and uneven) on PFJ stress during stair negotiation (both ascend and descend). We hypothesize that older adults will exhibit significant differences in PFJ stress waveform, but not magnitude compared to young adults during stair ascent and descent task, and all participants will increase magnitude of PFJ stress, but not change waveform on the challenging surfaces.

Methods

Participants

We recruited two cohorts, with 12 participants per cohort (Table 3.1). The first cohort consisted of young, healthy adults (between 18 and 25 years of age), with no history of musculoskeletal injury or disease. The second cohort was consisted of older adults (over 65 years of age), who have reported at least one accidental fall 12 months prior to testing. Any potential participant that self-reported: (1) a history of back or lower extremity injury or surgery, (2) current (in the past six months) pain or recent injury to the back or lower extremity and/or (3) any known neurological disorder were excluded. Participants in each cohort were matched by sex, height, and body mass index. Prior to testing, research approval was obtained from the local Institutional Review Board, and each participant provided written consent to participate.

Table 3.1 Mean (SD) subject demographics for each cohort (young and older adults).

	N	Age (yrs)	Height (m)	Weight (kg)	Walking Speed (m/s)
Young Adults	12 (f = 6)	21.08 (1.93)	1.75 (0.10)	68.91 (16.86)	1.06 (0.83)
Older Adults	12 (f = 6)	69.92 (3.15)	1.73 (0.13)	75.05 (17.71)	1.04 (0.17)
p-value	-	<0.001	0.674	0.394	0.720

Experimental Protocol

Each participant performed one orientation and one test session. The orientation sessions lasted approximately 30 minutes, while the test session lasted approximately four hours. The orientation and test session were separated by at least 24 hours to minimize effect of fatigue.

Orientation Session

The orientation session was used to collect participant demographic and strength data, and to familiarize each participant with the test procedures. During orientation, participant demographic information, including height (m), weight (kg), and age (years) as well as foot dominance was recorded. Foot dominance was determined by asking the participant which foot they would prefer to kick a ball.[115] Each participant also had dominant lower limb strength recorded on an isokinetic dynamometer (HUMAC NORM, CSMI, Stoughton, MA, USA). Specifically, maximal isometric hip and knee flexion and extension, and ankle dorsi- and plantar-flexion strength were recorded. For hip flexion and extension, participants stood with the hip flexed at 15 degrees. For knee flexion and extension, participants were seated with the hip and knee secured at 90 and 60 degrees, respectively.[116] For ankle dorsi and plantar-flexion, participants laid prone with the ankle neutral (0 degrees of plantarflexion).[117] For each movement, participants performed three maximal 5 second isometric contractions, with 15 seconds of rest between each contraction. Participants were given a minimum of 40 seconds of rest between movements.[118] Maximum torque produced during each contraction was recorded. During orientation, participants also were afforded the opportunity to familiarize themselves with the study activities. Each participant was required to give verbal confirmation that they can perform all study tasks at the conclusion of the orientation.

Biomechanical Testing

During each test session, participants completed four activities (walk, pivot, stair ascent, and stair descent) across three different surfaces (normal, slick, and uneven). Throughout testing, participants were outfitted with black spandex shorts and shirt, and wore their own broken-in tennis shoes. In order to prevent falls, participants were required to wear a safety harness connected to an overhead gantry that spans the entire motion capture volume during each study task (Figure 3.1). To avoid bias and confounding data, a Latin Square Design was used to randomly assign the activity and surface order prior to testing (Table 3.2 and Table 3.3).

Figure 3.1 Weight supporting gantry used for stair ascent and stair descent task

Table 3.2 Latin Square Design used for randomization of the testing order for each activity.

	Task 1	**Task 2**	**Task 3**	**Task 4**
Order 1	Pivot	Stair Descent	Walk	Stair Ascent
Order 2	Stair Ascent	Pivot	Stair Descent	Walk
Order 3	Walk	Stair Ascent	Pivot	Stair Descent
Order 4	Stair Descent	Walk	Stair Ascent	Pivot

Table 3.3 Latin Square Design used for randomization of the testing order for each surface.

	Surface 1	**Surface 2**	**Surface 3**
Order 1	Normal	Uneven	Slick
Order 2	Slick	Normal	Uneven
Order 3	Uneven	Slick	Normal

During each test session, participants had three-dimensional (3D) lower limb (hip, knee and ankle) biomechanical data recorded during each study task. Ground reaction force (GRF) data (2400 Hz) was recorded with one inground force platform (AMTI OR6 Series, Advanced Mechanical Technology Inc., Watertown, MA), while ten high-speed (240 Hz) optical cameras (Vantage, Vicon Motion Systems, LTD, Oxford, UK) recorded lower limb motion data.

For this study, only the stair ascent and descent tasks were analyzed and thus, are the only tasks described below. For each stair negotiation task, the participant walked at a predetermined, self-selected speed to either ascend and descend two stairs (18.5cm rise) fixated on top of the force platform (Figure 3.2). Staircase height was determined based on the requirements of the 2021 International Residential Code that states stairs should not exceed 7.75 inches (19.7 cm).[119] To determine a participants' self-selected speed, they performed a walking task through two sets of infrared timing gates (TracTronix TF100, TracTronix Wireless Timing Systems, Lenexa, KS) within the motion capture volume (about 10 meters) placed 1.8 meters apart. The walking task consisted of level-walking in which participants were asked to walk at a comfortable speed through the timing gates five times. Then, their self-selected speed was calculated as the average of those five trials. For the stair ascent task, participants walked through the level motion capture space, placed their dominant limb on the target (first) step (18.5 cm rise), before ascending to the second step. For the stair descent task, participants started atop the second step, and descended the stairs by placing their dominant limb on the target (first) stair and then walked through the motion capture volume at the participant-selected speed.

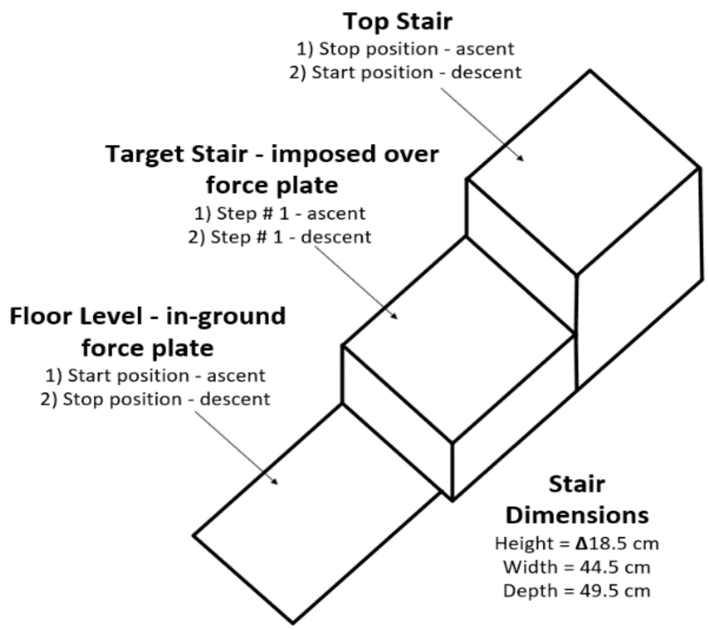

Figure 3.2 Staircase model used in motion capture space for stair ascent and descent task

Participants performed each stair negotiation task on three different stair surfaces (1: normal, 2: slick, and 3: uneven) (Figure 3.3). Each surface was fixed to the top of the target and top stair. The normal surface consisted of a flat, painted wood panel. The slick surface consisted of a wood panel covered by a smooth, plastic material that, when combined with slick booties each participant was required to wear, produced a coefficient of static friction between the shoes and surface (0.19) comparable to ice (0.10) (Figure 3.4).[120] The uneven surface consisted of a wood panel composed of nine painted wooden blocks of differing heights. Each participant performed three successful trials across each surface for both stair negotiation (both ascend and descend) tasks. Trials were considered successful if the participant walked within ± 5 % of their pre-determined speed, only contacted the target stair with their dominant limb, and did not slip or trip during the trial.

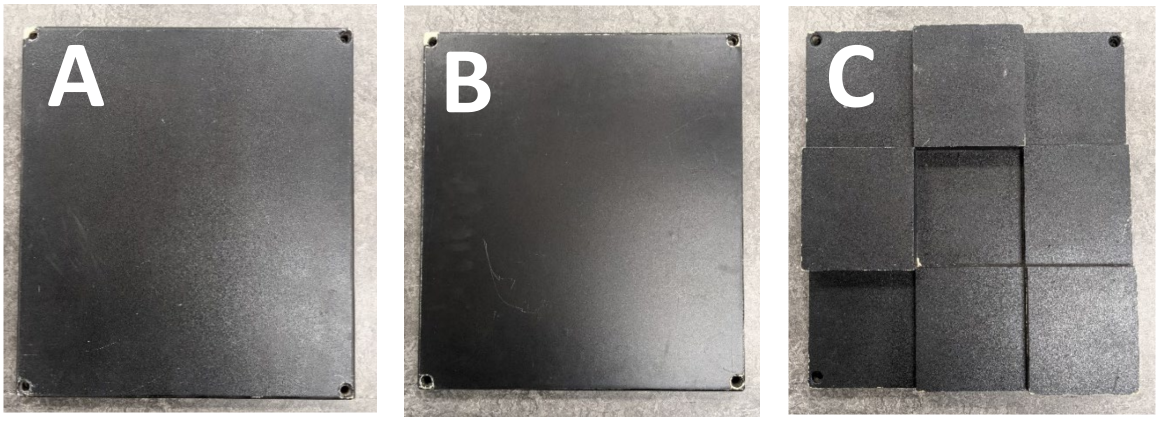

Figure 3.3 Normal (A), Slick (B) and Uneven (C) for stair ascent and descent task

Figure 3.4 Slick booties outfitted over participant shoes with holes for marker visibility during slick surface stair negotiation

<u>Biomechanical Analysis</u>

During each trial, lower limb biomechanical data was quantified from 3D coordinates of 50 retro-reflective and four virtual markers (Table 2.4). Each reflective marker was attached with double sided tape and secured using elastic tape (Cover-Roll Stretch, BSN Medical GmbH, Hamburg, Germany) over a specific body landmark. Each virtual marker was digitized in the global coordinate system using a Davis Digitizing Pointer (C-Motion Inc., Rockville, MD). After all markers were secured or digitized, each participant stood in anatomical position for a static recording. The static recording was used to create a kinematic model that consists of 8 segments (trunk, pelvis, and bilateral thigh, shank, and foot) and 27 degrees of freedom in Visual 3D (v6, C-Motion, Inc, Germantown, MD, USA). Each model segment was assigned a local coordinate system and three orthogonal axes (x, y, and z). For the trunk, the origin was defined at the intersection of the midpoint of the acromion processes and the seventh cervical vertebrae and sternum jugular notch, and assigned a local coordinate system with three degrees of freedom. For the pelvis, the origin was defined as halfway between the right and left anterior superior iliac spines, and assigned a local coordinate system with three rotational and three translational degrees of freedom. For the thigh, a functional hip joint center was determined in accordance with Schwartz and Rozumalski[121] and set as the origin, and assigned a local coordinate system with three degrees of freedom. The shank and foot had local coordinate systems with three degrees of freedom, and knee and ankle joint centers set as the segment origin and defined as the midpoint between medial and lateral femoral epicondyle and medial and lateral malleoli in accordance to Grood and Suntay, and Wu, respectively.[122,123]

Table 3.4 Retro-reflective and Virtual Whole-body Marker Set

	Markers
Trunk	*Acromion process,* **jugular notch, xiphoid process, V7 vertebrae, T12 vertebrae**
Pelvis	*Anterior-superior iliac spines, posterior-superior iliac spines,* and iliac crests
Thigh	Greater trochanter, distal thigh, *medial* and *lateral femoral epicondyles*
Shank	Tibial tuberosity, lateral fibula, distal tibia, *medial* and *lateral malleoli*
Foot	Posterior heel, *first* and *fifth metatarsal heads*

Note: *Italic* indicates calibration markers. ***Bold*** indicates virtual markers.

For each trial, the synchronous GRF and marker trajectory data were low pass filtered using a fourth-order Butterworth filter with cut-off frequency of 12 Hz. Then, filtered marker trajectories were processed in Visual 3D, using a joint coordinate system approach, to calculate knee rotations expressed with respect to a participant's static pose.[6] The filtered kinematic and GRF data were processed to obtain 3D knee forces and moments using standard inverse-dynamics analyses, and segment inertial properties defined according to Dempster.[124,125] All biomechanical data was normalized from 0% to 100% of stance phase and resampled to 1% increments (N=101). Stance phase was identified as heel strike to toe-off and defined as the moment when GRF first exceeded and fell below 10 N, respectively.

Custom MATLAB (R2021b, Mathworks, Natick, MA) code was used to calculate stance phase PFJ stress, as a function of knee flexion joint angle and knee extension joint moment based on a two-dimensional biomechanical model according to Brechter and Powers (2002).[20] Specifically, the model inputs are knee joint flexion angle and extension moment obtained from data collection, and quadriceps lever arm, a constant (k), and PFJ contact area obtained from previous experimental data.[126-129] First, the quadriceps effective lever arm (LA; fit ($r^2 = 0.99$) to data of van Eijden et al) and quadriceps force (QF) were determined using Equations 1 and 2:[126]

Equation 1 $$LA(x) = (8.0E\text{-}05x^3 - 1.3E\text{-}02x^2 + 2.8E\text{-}01x + 0.046);$$

Equation 2 $$QF(x) = M_{EXT}(x) / LA(x);$$

where: LA = effective quadriceps lever arm (m), x = flexion angle (deg),

QF = quadriceps force (N), and M_{EXT} = knee extension moment (N*m).

Next, the PFJ reaction force was estimated from a constant (k; fit ($r^2 = 0.99$) to data of van Eijden and colleagues[127,128]) that represents the ratio of patellofemoral compression force and the quadriceps force as a function of knee flexion angle, using Equations 3 and 4:

Equation 3

$$k(x) = (4.62E\text{-}01 + 1.47E\text{-}03x - 3.84E\text{-}05x^2) / (1 - 1.62E\text{-}02x + 1.55E\text{-}04x^2 - 6.98E\text{-}07x^3);$$

Equation 4 $PFJRF(x) = k(x) * QF(x);$

where: k = constant (N/N), x = knee flexion angle (deg),

PFJRF = PFJ reaction force (N), and QF = quadriceps force (N).

Finally, contact area was calculated as a function of knee angle and PFJ stress determined as the ratio of PFJ reaction force and contact area (PFJCA: fit ($r^2 = 0.99$) to data of Connolly et al[129]) using Equations 5 and 6:

Equation 5 $PFJCA(x) = (7.81E\text{-}02x^2 + 6.763E\text{-}01x + 151.75);$

Equation 6 $PFJ\ Stress(x) = PFJRF(x) / PFJCA(x);$

where: PFJCA = PFJ contact area (mm^2), x = knee flexion angle (deg),

PFJ Stress = PFJ stress (N/mm^2 or MPa),

PFJRF = PFJ reaction force (N), and

PFJCA = PFJ contact area (mm^2).

Statistical Analysis

Predefined knee biomechanics related to PFJ stress were submitted to statistical analysis. Specifically, the discrete dependent variables included peak of stance (PS, 0-

100%) knee flexion joint angle and moment, peak and impulse PFJ reaction force, and peak, time integral and time of peak PFJ reaction force and stress as well as stance time and average and range (peak minus minimum) PFJ contact area. Each dependent variable was averaged across the three successful trials to create a participant-based mean. Then, each participant-based mean was submitted to a mixed-model ANOVA to test the main effects of and interaction between age (young vs old) and surface (normal, slick, and uneven). Significant interactions were submitted to simple effects analysis and a Bonferroni correction will be used for significant pairwise comparisons.[130] Alpha was set to a *priori* at $p < 0.05$. All statistical analysis was performed using SPSS v25 software (IMB, Armonk, NY).

Statistical Parametric Mapping (SPM) (see Appendix B for further specifics regarding SPM), a technique for statistically understanding 1-dimensional (1-D) temporal/spatial regions where significant differences may occur, was used to compare the PFJ stress waveform between groups and conditions. Specifically, a SPM two-way ANOVA with one repeated measure was used to determine main effects of and interaction between age and surface. If the scalar output statistic (SPM{F}) crossed the critical threshold for statistical significance at any time point, a supra-threshold was defined and the associated p-values were calculated using Random Field Theory.[131,132] If a supra-threshold cluster was found, follow-up SPM t-tests (SPM{t}) ($p < 0.05$) were performed to identify changes within each main effect or interaction. All SPM analysis was conducted in a custom MATLAB code implementing functions from the open-source spm1d package (www.spm1d.org).

Results

There was a significant difference in age ($p < 0.001$), but not height ($p = 0.674$), weight ($p = 0.394$), or self-selected walking speed ($p = 0.720$) between young and older adults (Table 3.1).

<u>Stair Ascent</u>

There was a significant age by surface interaction for time of peak PFJ stress ($p = 0.041$) (Figure 3.5 and Table 3.5). Older adults exhibited significantly later peak PFJ stress on the uneven ($p = 0.008$), but not normal or slick surfaces ($p > 0.05$) compared to young adults. Older adults peak PFJ stress was later on the uneven surface compared to normal and slick surfaces (both: $p = 0.014$), while young adults exhibited no significant difference in time of peak PFJ stress on any surface ($p > 0.05$).

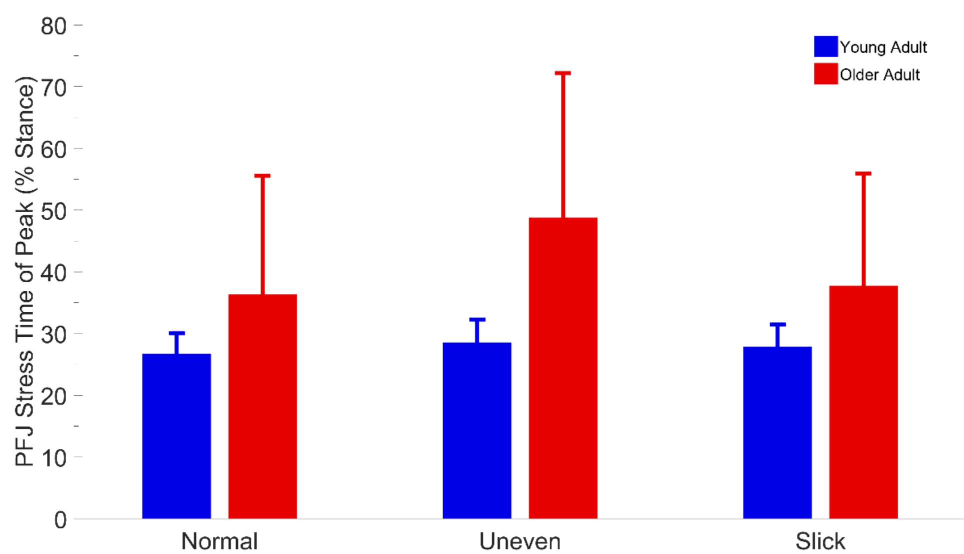

Figure 3.5 Mean ± SD for PFJ stress time of peak between young and older adults on each surface during stair ascent.

Older adults exhibited greater PFJ stress-time integral (p = 0.004), PFJ reaction force impulse (p = 0.030), and later peak PFJ stress (p = 0.024) than young adults (Table 3.5). But, no significant difference for peak PFJ stress and reaction force, PFJ contact area (both range and mean), or peak knee flexion angle and extension moment (p > 0.05) was evident between cohorts.

Table 3.5 Mean (SD) for PFJ measures during stair ascent by age and surface.

	Normal		**Uneven**		**Slick**	
	Young	**Older**	**Young**	**Older**	**Young**	**Older**
Peak PFJ Stress (MPa)	3.93 (1.26)	4.29 (1.51)	3.77 (1.29)	4.39 (1.44)	3.81 (1.13)	4.19 (1.48)
PFJ Stress-time Integral (MPa* %stance)*†	116.93 (51.00)	203.34 (72.82)	140.40 (65.50)	226.13 (71.06)	120.12 (60.04)	199.95 (70.66)
PFJ Stress Time of Peak (%stance)*#	26.75 (3.33)	36.33 (19.27)	28.58 (3.73)	48.75 (23.45)	27.92 (3.55)	37.75 (18.22)
Peak PFJ Reaction Force (N)†	1363.20 (419.82)	1324.98 (443.42)	1470.91 (503.09)	1385.51 (450.06)	1315.82 (352.62)	1283.64 (420.01)
PFJ Reaction Force Impulse (N*%stance)*†	37565.62 (14521.02)	55893.07 (21385.67)	46296.51 (19898.42)	64343.53 (21.060.58)	38201.70 (16602.49)	54171.18 (19651.37)
PFJ Contact Area Range (mm^2)†	227.60 (35.37)	213.02 (54.74)	288.15 (45.22)	262.15 (48.94)	220.77 (35.18)	202.12 (56.65)
PFJ Contact Area Mean (mm^2)†	249.57 (32.27)	256.20 (55.89)	282.68 (35.89)	281.79 (47.06)	256.11 (29.47)	255.13 (50.87)

* Denotes a significant (p < 0.05) main effect of age
† Denotes a significant (p < 0.05) main effect of surface
# Denotes a significant (p < 0.05) significant interaction between age and surface

Surface impacted both PFJ measures and knee flexion biomechanics. Specifically, there was a main effect of surface for every PFJ measure (all: $p < 0.05$), except peak PFJ stress ($p = 0.288$) (Table 3.5). Participants exhibited greater PFJ stress-time integral, PFJ reaction force impulse, and PFJ contact area (both mean and range) on the uneven compared to the normal and slick surfaces (all: $p < 0.010$). Peak PFJ reaction force was also greater on uneven compared to slick ($p = 0.006$), but not normal surface ($p = 0.108$). After correcting for type I error, no significant difference in time of peak PFJ stress was observed between any surface ($p > 0.05$). In addition, a main effect of surface was observed for peak knee flexion angle ($p < 0.001$) and extension moment ($p = 0.002$) (Table 3.6). Participants exhibited greater peak knee flexion angle on the uneven compared to normal surface ($p < 0.001$) (Figure 3.6), and greater peak knee extension moment on the uneven compared to the slick surface ($p = 0.004$) (Figure 3.7).

Table 3.6 Mean (SD) for knee flexion biomechanics during stair ascent by age and surface.

	Normal		**Uneven**		**Slick**	
	Young	**Older**	**Young**		**Young**	**Older**
Peak Knee Flexion Angle (deg)†	51.82 (4.81)	49.88 (8.71)	58.71 (5.20)	55.63 (6.89)	51.37 (5.13)	48.65 (8.69)
Peak Knee Extension Moment (Nm/kg*m)†	0.77 (0.13)	0.70 (0.15)	0.81 (0.14)	0.73 (0.12)	0.75 (0.13)	0.68 (0.12)

† Denotes a significant ($p < 0.05$) main effect of surface

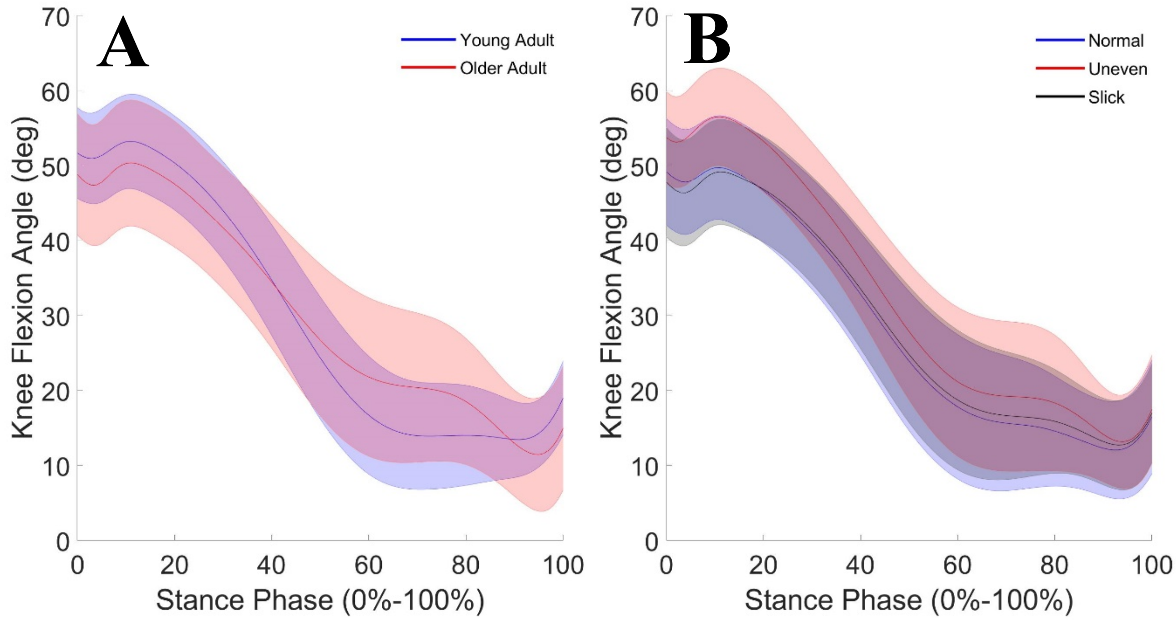

Figure 3.6 Mean ± SD for knee flexion angle by age (A) and surface (B) during stair ascent.

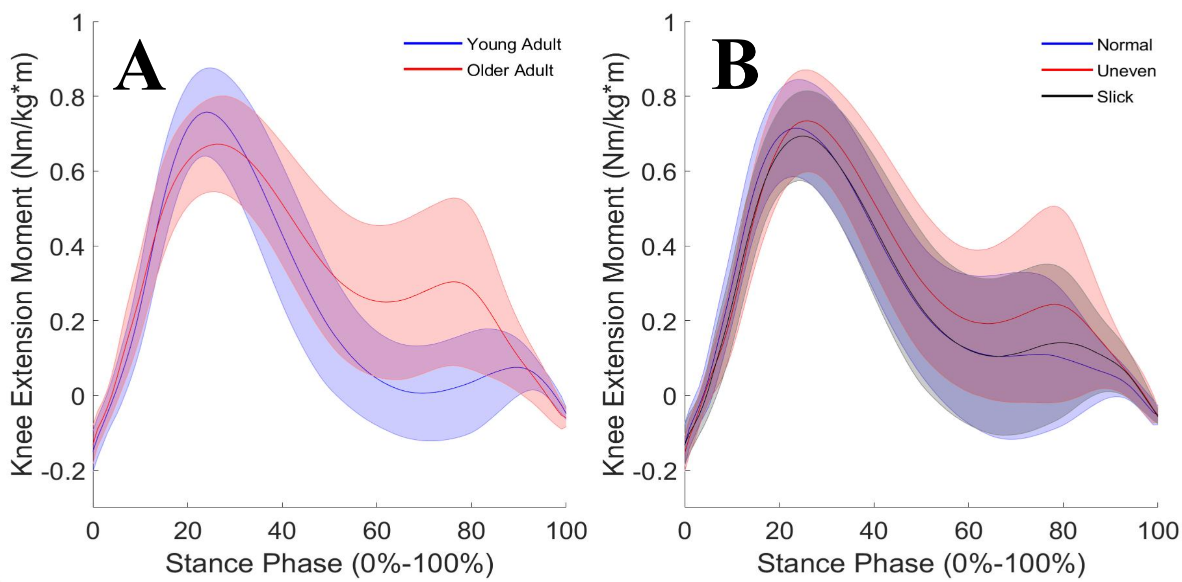

Figure 3.7 Mean ± SD for knee extension moment by age (A) and surface (B) during stair ascent.

SPM analysis revealed a main effect of age ($p < 0.001$) (Figure 3.4) and surface from 8 to 26% and 44 to 93% of stance (both $p < 0.001$) (Figure 3.5) for the PFJ stress waveform. Specifically, older adults exhibited greater PFJ stress from 56 to 84% of stance compared to young adults ($p < 0.001$). On the uneven surface, participants exhibited smaller PFJ stress from 8 to 25% of stance and significantly greater PFJ stress from 57 to 90% of stance compared to normal surface (both: $p < 0.001$) (Figure 3.9C), and greater PFJ stress from 49 to 77% of stance compared to slick surface ($p = 0.002$) (Figure 3.9D).

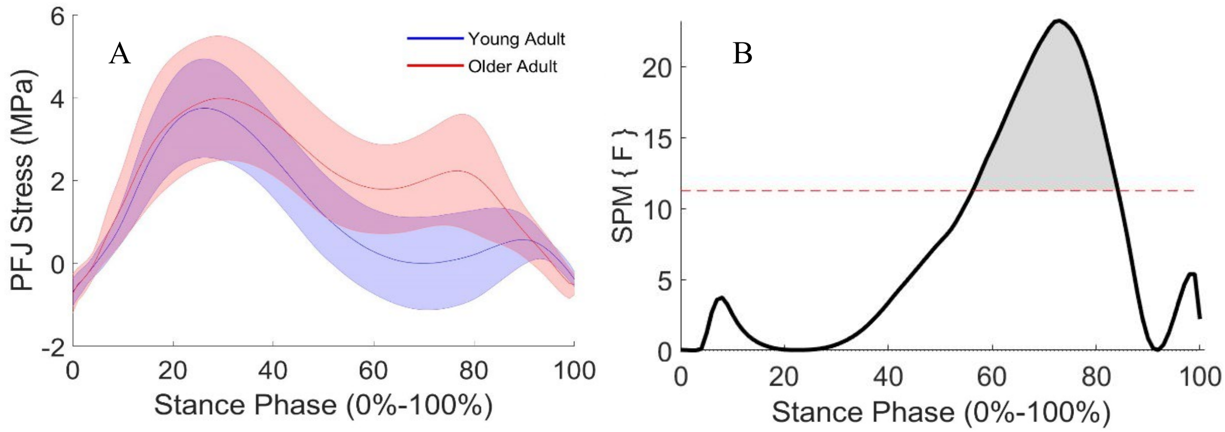

Figure 3.8 Stair ascent stance phase (0-100%) PFJ stress waveform by age (A). SPM analysis revealed significant main effect waveform differences by age (B).

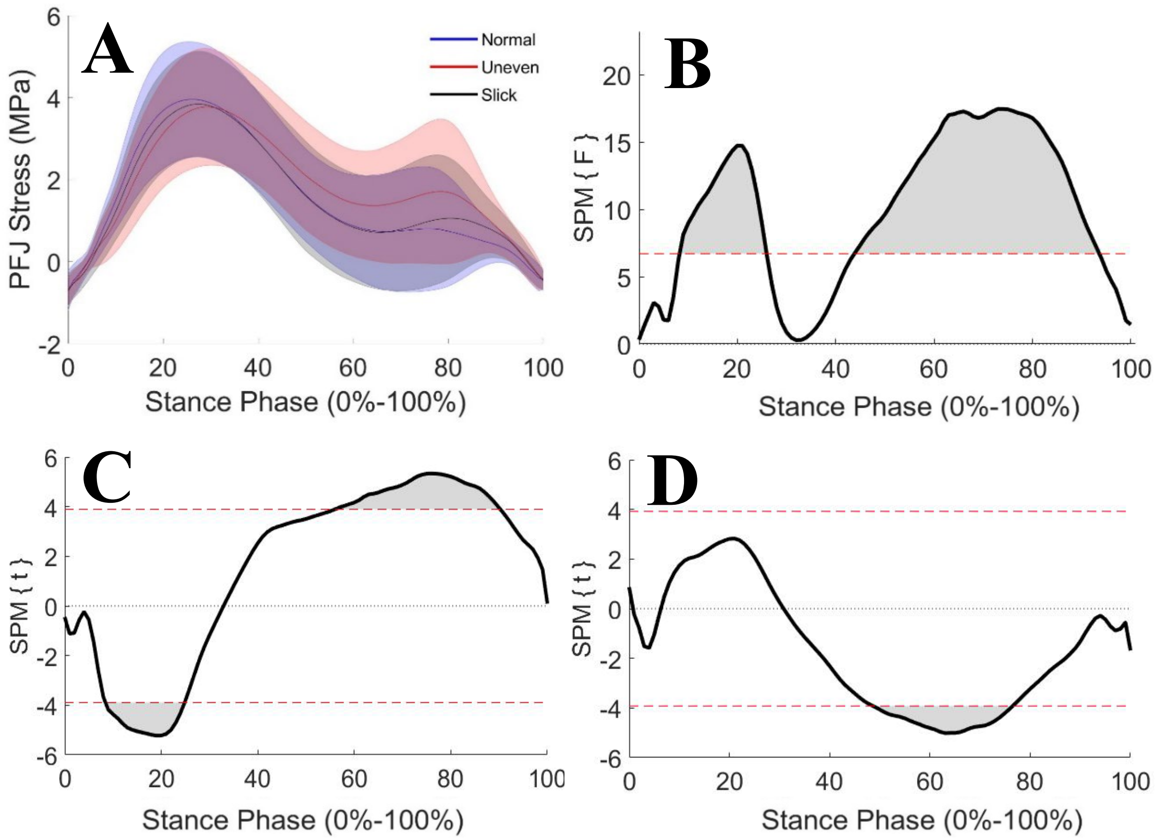

Figure 3.9 Stair ascent stance phase (0-100%) PFJ stress waveform by surface (A). SPM analysis revealed significant main effect waveform differences by surface (B). Post-hoc analysis revealed significant differences between uneven and normal (C) and uneven and slick (D) surfaces.

Stair Descent

Older adults exhibited smaller PFJ contact area range (p = 0.018) (Table 3.7) and peak knee flexion angle (p = 0.022) (Figure 3.10 and Table 3.8) than young adults. No significant difference (p > 0.05) between young and older adults was observed for any other PFJ measure or peak knee extension moment.

Table 3.7 Mean (SD) for PFJ measures during stair descent by age and surface.

	Normal		Uneven		Slick	
	Young	**Older**	**Young**	**Older**	**Young**	**Older**
Peak PFJ Stress (MPa)	3.78 (1.02)	4.34 (1.66)	3.69 (1.01)	4.30 (1.78)	3.92 (1.25)	4.18 (1.66)
PFJ Stress-time Integral (MPa * % stance)†	170.86 (40.70)	212.63 (92.56)	184.84 (65.87)	241.99 (127.17)	166.24 (53.32)	210.40 (90.60)
PFJ Stress Time of Peak (% stance)	46.50 (29.65)	60.75 (28.08)	55.67 (29.71)	61.00 (25.13)	48.33 (29.68)	74.08 (17.20)
Peak PFJ Reaction Force (N)†	1273.19 (403.29)	1378.96 (438.91)	1423.36 (352.14)	1559.91 (559.79)	1277.76 (425.66)	1467.30 (495.64)
PFJ Reaction Force Impulse (N * % stance)†	51900.02 (11113.17)	59261.71 (21391.52)	58960.37 (18309.88)	71587.41 (31980.17)	51780.27 (16929.19)	60881.05 (22159.33)
PFJ Contact Area Range (mm^2)*	582.58 (65.02)	503.75 (65.26)	590.85 (104.67)	500.34 (92.02)	554.304 (95.85)	488.91 (58.50)
PFJ Contact Area Mean (mm^2)†	314.65 (80.61)	277.92 (45.47)	277.92 (45.47)	299.21 (53.52)	305.50 (56.34)	278.02 (36.62)

* Denotes a significant (p < 0.05) main effect of surface
† Denotes a significant (p < 0.05) main effect of surface

Table 3.8 Mean (SD) for knee flexion biomechanics during stair descent by age and surface.

	Normal		Uneven		Slick	
	Young	**Older**	**Young**	**Older**	**Young**	**Older**
Peak Knee Flexion Angle (deg)*	82.57 (5.04)	76.32 (5.67)	83.09 (7.96)	76.28 (7.45)	80.28 (7.36)	75.17 (5.22)
Peak Knee Extension Moment (Nm/kg*m)†	0.72 (0.15)	277.92 (45.47)	0.81 (0.16)	0.80 (0.17)	0.73 (0.14)	0.76 (0.13)

* Denotes a significant ($p < 0.05$) main effect of surface
† Denotes a significant ($p < 0.05$) main effect of surface

Surface impacted PFJ stress-time integral, peak and impulse PFJ reaction force, and mean PFJ contact area (all: $p < 0.006$) (Table 3.7) as well as peak knee extension moment ($p = 0.003$) (Figure 3.11). On the uneven surface, participants exhibited greater peak and impulse of PFJ reaction force and mean PFJ contact area compared to the normal surface (all: $p < 0.05$), and greater PFJ stress-time integral, PFJ reaction force impulse, and mean PFJ contact area compared to slick surface (all: $p < 0.05$). Peak knee extension moment was greater on the uneven surface compared to normal ($p = 0.014$) and slick ($p = 0.029$) surfaces.

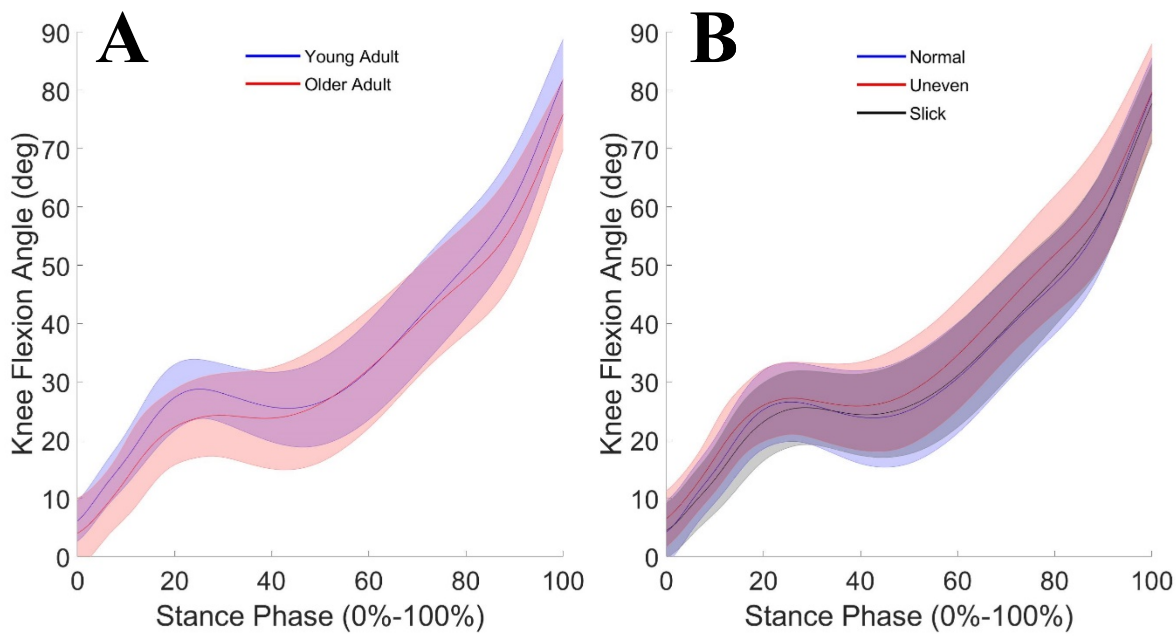

Figure 3.10 Mean ± SD for knee flexion angle by age (A) and surface (B) during stair descent.

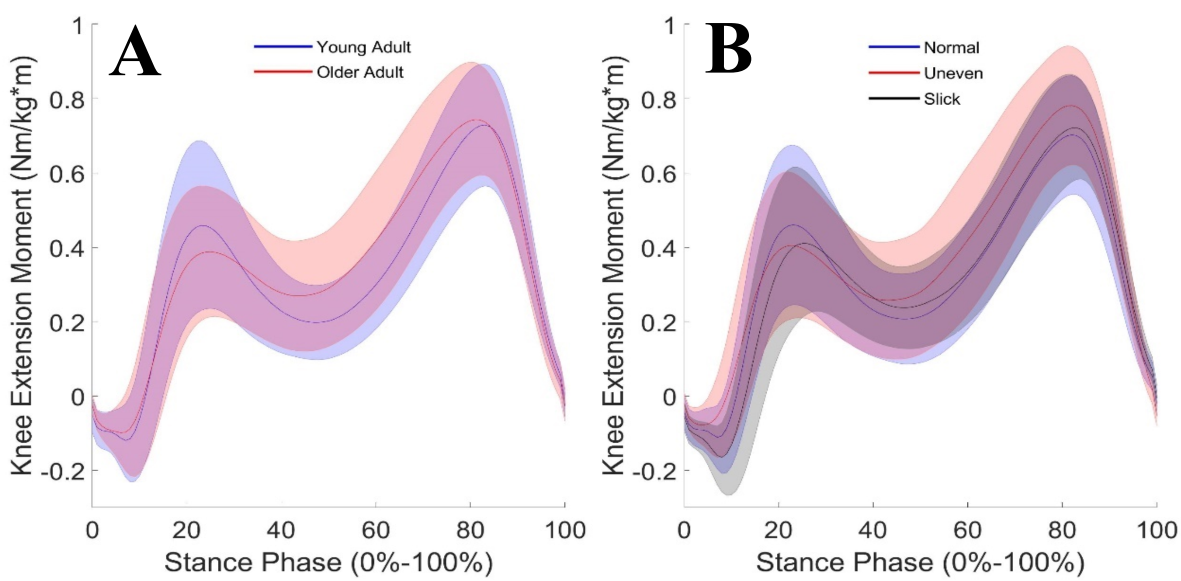

Figure 3.11 Mean ± SD for knee extension moment by age (A) and surface (B) during stair descent.

SPM analysis of the PFJ stress waveform revealed a main effect of surface from 0 to 2% ($p = 0.046$), 3 to 18% ($p < 0.001$), 53 to 69% ($p < 0.001$), and 93 to 100% ($p = 0.015$) of stance (Figure 3.13). On the uneven surface, participants exhibited greater PFJ stress from 5 to 16% stance ($p < 0.001$), but smaller PFJ stress from 98 to 100% of stance compared to the slick ($p = 0.013$) surface, and greater PFJ stress from about 99.5 to 100% of stance compared to normal ($p = 0.017$) surface. On the slick surface, participants exhibited smaller PFJ stress from 5 to 18% of stance ($p < 0.001$), but greater PFJ stress from 92 to 99% of stance compared to the normal ($p = 0.002$) surface.

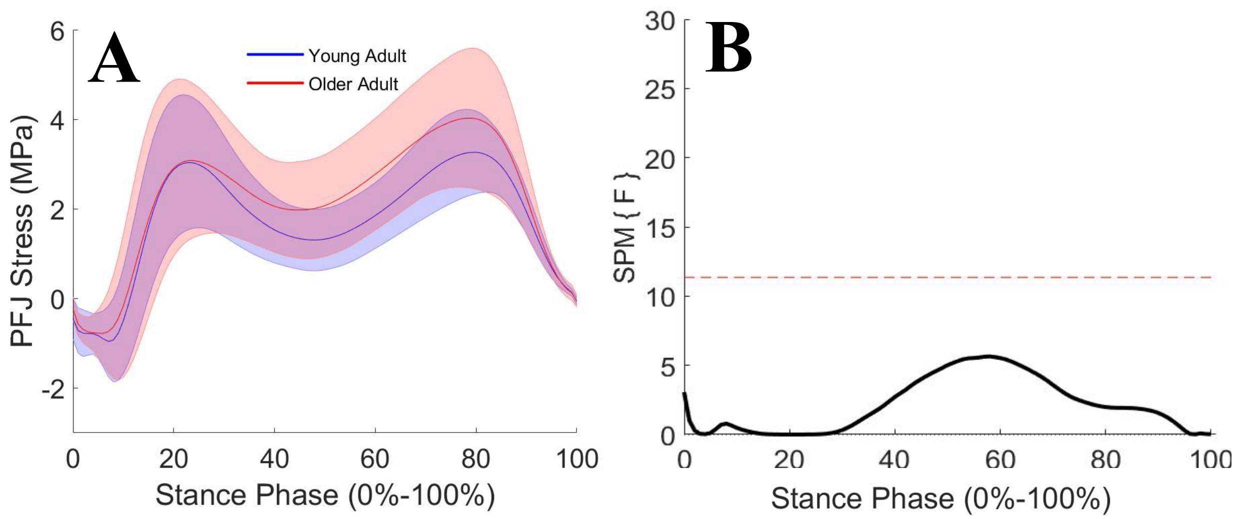

Figure 3.12 Stair descent stance phase (0-100%) PFJ stress waveform by age (A). SPM analysis revealed significant main effect waveform differences by age (B).

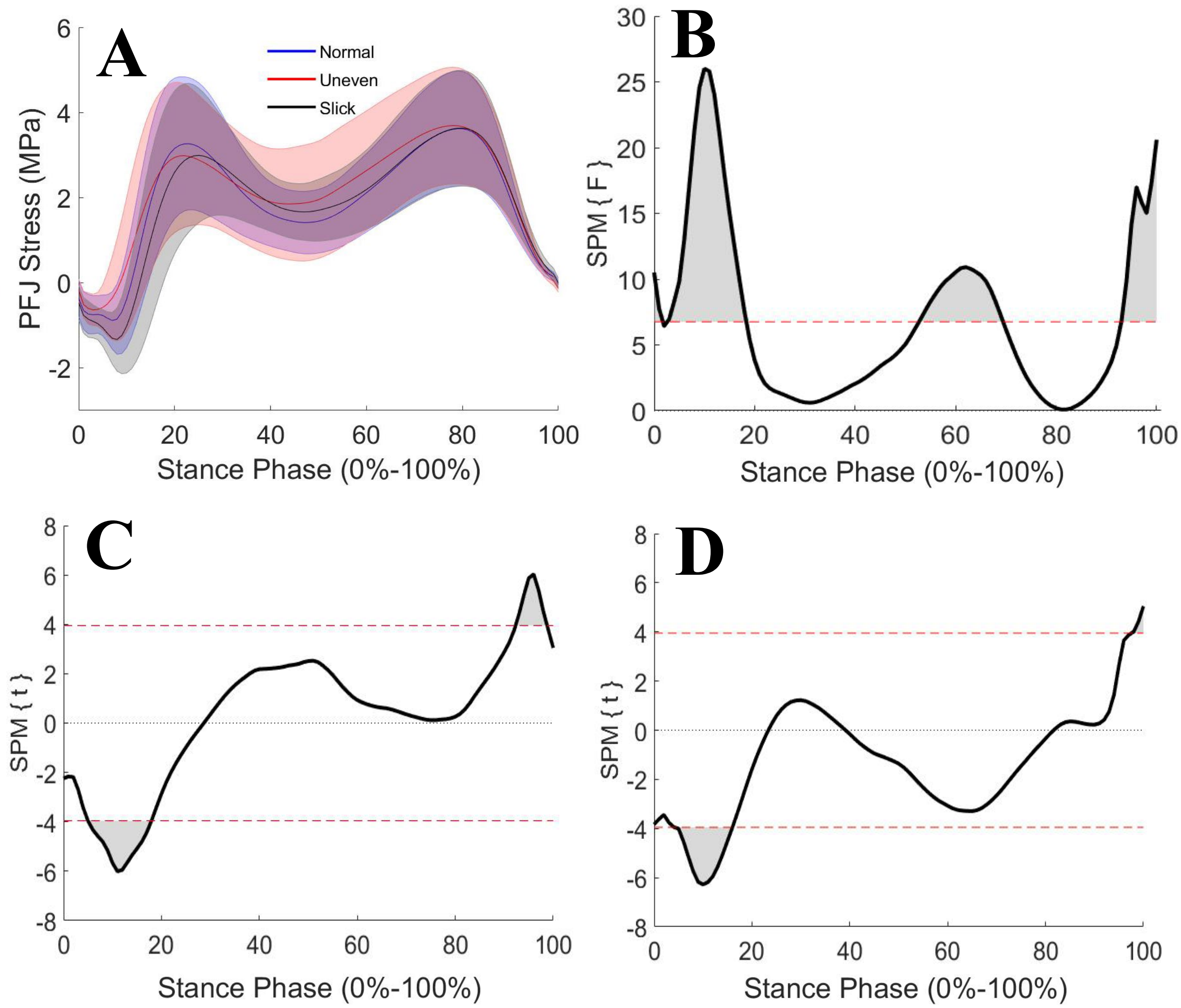

Figure 3.13 Stair descent stance phase (0-100%) PFJ stress waveform by surface (A). SPM analysis revealed significant main effect waveform differences by surface (B). Post-hoc analysis revealed significant differences between slick and normal (C) and slick and uneven (D) surfaces.

Discussion

This study examined PFJ stress magnitude and waveform for young and older adults negotiating stairs with challenging surfaces. In partial agreement with our hypothesis, both age and surface impacted PFJ stress waveform, but not magnitude during the stair ascent task; whereas, only surface impacted the PFJ stress waveform during the stair descent.

Older adults exhibited waveform differences in PFJ stress that may increase their risk of developing PFJ pain. Specifically, during the stair ascent, older adults exhibited greater PFJ stress from 56 to 84% of stance (i.e., mid to late stance), which may stem from the 67% increase in PFJ stress-time integral and significantly later peak PFJ stress they exhibited compared to their younger counterparts. The larger, later PFJ stress may increase their risk of developing PFJ pain, as it is reportedly exhibited by individuals with confirmed PFJ pain and reportedly increases damage to the joint's articular cartilage[20,133-136], which may predispose older adults to knee pain development.[137,138] Although non-significant, older adult's large 12% increase in peak PFJ stress compared to young adults during the stair ascent may further load their articular cartilage. However, the reason older adults exhibited differences in the PFJ stress waveform, and not peak PFJ stress, is not immediately evident. Considering peak PFJ stress is purportedly related to peak knee flexion biomechanics[20], the fact no age dimorphism in peak knee flexion angle and moment were currently observed during the stair ascent may contribute to the insignificant difference in peak PFJ stress. Future work, nonetheless, is warranted to determine if waveform differences in knee flexion biomechanics contribute to magnitude and waveform differences in PFJ stress.

Similar significant age differences in PFJ stress were not observed during the stair descent. Contrary to our hypothesis, older adults exhibited a non-significant, albeit large, 13% and 27% increase in peak PFJ stress and stress-time integral during the stair descent. While the reason the large increases in PFJ stress for older adults did not reach statistical significance is not immediately evident it may be attributed to the large variability exhibited by the current participants, but particularly the older adults, during the stair descent. Specifically, older adults' coefficient of variation (or measure of relative variability[139]) for peak PFJ stress and stress-time integral was 40 and 47%, and approximately 10% greater than the young adults. In general, older adults exhibit more variable gait due to age-related alterations in musculoskeletal, cognitive, and sensorimotor function than their younger counterparts.[140-143] The current older adults age-related losses of quadriceps strength (Appendix A and Table A.1) may contribute to both the large PFJ stress variability and specific alterations in knee biomechanics they exhibited during the stair ascent. In particular, the older adults exhibited a 7% reduction in peak knee flexion angle and a 14% decrease in the range of PFJ contact area during the stair descent. Both the reduction in knee flexion angle and PFJ contact area are biomechanical alterations reported to increase PFJ stress, and articular cartilage damage.[144,145] These strategies may be adopted by the weaker older adults to prevent limb collapse when descending stairs, as a more extended limb is a biomechanical adaptation to prevent overwhelming the quadriceps musculature and ensuing limb collapse.[146]

The challenging surfaces, particularly uneven, impacted PFJ stress and knee flexion biomechanics during stair ascent. Interestingly, on the uneven surface, participants decreased PFJ stress during weight acceptance (8 to 25% of stance), but increased stress during terminal stance (57 to 90% of stance) when ascending stairs. Both changes may be attributed to specific knee biomechanics adopted by the participants during each gait phase. In particular, participants exhibited 12% greater peak knee flexion as well as between 12% and 25% larger PFJ contact area (both mean and range) on the uneven compared to normal surface. The larger knee flexion posture may increase the mechanical advantage of the quadriceps and require smaller muscular force to prevent limb collapse during stair ascent, while a larger PFJ contact area would disperse the quadriceps force and PFJ stress across more of the knee joint's surface, decreasing likelihood of joint damage.[144,145,147] Both adaptations may aid the participants ability to limit magnitude of PFJ stress, and contributed to the observed reduction during weight acceptance. Conversely, participants exhibited a 14% increase in PFJ stress-time integral and 18% increase in PFJ reaction force impulse on the uneven compared to normal surface. The larger PFJ stress-integral and PFJ reaction force may contribute to the elevated PFJ stress during terminal stance as well as likelihood of PFJ pain development, particularly for the older adults. Further, on the uneven surface, older adults exhibited a 70% later peak PFJ stress compared to young adults, and 34% later peak stress compared to normal and slick surfaces. The older adults delayed peak PFJ stress may contribute to their elevated PFJ stress in terminal stance, and apply greater force to PFJ articular cartilage during those gait phases.[147] Yet, future work is needed to determine if the delayed peak PFJ stress exhibited by older adults on the uneven surface further increases their risk of developing PFJ pain.

The challenging surfaces also impacted PFJ stress during stair descent, and potential to damage the joint's articular cartilage. Although no significant difference in peak PFJ stress was observed when descending stairs on the challenging surfaces, both the uneven and slick surface elicited changes in the PFJ stress waveform. When descending on the uneven surface, participants exhibited greater PFJ stress in early stance (5 to 16%) and smaller stress in late stance (98 to 100%) compared to the slick surface. The alterations of PFJ stress waveform on the uneven compared to slick surface may result from large increases (between 8% and 16%) of the PFJ stress-time integral, reaction force, and contact area. Interestingly, on the uneven surface, participants also exhibited a substantial increase in peak and impulse PFJ reaction force (12% and 17%, respectively) compared to the normal surface, but no significant difference in either peak or waveform of PFJ stress was observed with the normal surface. Conversely, on the slick surface participants exhibited smaller PFJ stress in early stance (5 to 18%) and greater stress in late stance (92 to 99%) compared to the normal surface, but no significance in discrete PFJ stress variables were observed with the normal surface. The large increases in PFJ stress-time integral and reaction force observed when navigating the challenging surfaces during the stair descent may load the joint's articular cartilage, leading to degradation and subsequent pain.[133-138] However, further study is needed to determine the effect of greater stress-time integral on tissue damage and the development of PFJ pain.

This study may be limited by the PFJ stress calculation. The current PFJ stress model may underestimate PFJ reaction force and subsequent stress, as the model does not account for hamstring muscle force and estimates PFJ contact area as a function of knee flexion angle. Considering, van Eijden et al. and Connolly et al. reported r^2 values of 0.99 for the predicted PFJ reaction force and estimated PFJ contact area with cadaveric and radiographically-derived measures, respectively, we are confident our PFJ stress measures accurately represent PFJ loading.[126-129] Further, the chosen challenging surfaces may be a limitation. Although the coefficient of friction of the slick surface was comparable to ice (0.19 vs 0.10), it may not elicit similar a similar compensatory response as real ice, and the staggered wooden blocks of the uneven surface may be predictable and not imitate the randomness of real-world uneven terrain.

In conclusion, older adults are more likely to exhibit knee biomechanics related to PFJ pain development when navigating stairs, particularly late in stance and on the uneven surface. Older adults exhibited larger, later PFJ stress when ascending, but not descending the stairs compared to their younger counterparts. These large increases in PFJ stress may load the joint's articular cartilage and predispose the older adults to PFJ pain development. All participants, regardless of age, exhibited alterations in knee biomechanics that may lead to greater PFJ stress when negotiating stairs with slick and uneven surfaces.

CHAPTER FOUR: CONCLUSION

Introduction

The purpose of this study was to: (1) determine whether aging or (2) challenging surfaces led to significant increases in magnitude and waveform of PFJ stress during stair negotiation. Key findings partially support the hypotheses that older adults exhibited significant increases the PFJ waveform, particularly late in stance, during stair ascent but not descent. Yet, all participants altered knee biomechanics that may increase the risk of PFJ pain development when ascending and descending challenging stairs.

Key Findings

Older adults exhibited waveform differences in PFJ stress that may increase their risk of developing PFJ pain. Specifically, during the stair ascent older adults exhibited greater PFJ stress in terminal stance which may be attributed to the delayed peak PFJ stress and greater stress-time integral compared to their younger counterparts. Similarly, all participants ascending the uneven stair exhibited greater PFJ stress late in stance with a greater stress-time integral compared to both the normal and slick surfaces. These alterations may increase load placed on the PFJ articular cartilage when ascending stairs, increasing risk of PFJ pain development. During the stair descent, surface, but not age impacted the PFJ stress waveform. On the uneven surface, participants increased PFJ stress early in stance and decreased PFJ stress in mid to late stance; whereas, on the slick surface, participants decreased PFJ stress early in stance and increased stress late in stance. During the stair descent, challenging surface type led to different knee biomechanics alterations and magnitude of PFJ stress during early and late stance.

Significance

These findings support the precept that sagittal knee biomechanical alterations during stair negotiation may contribute to PFJ stress, and increase musculoskeletal injury and disease risk, particularly PFJ pain. Specifically, this study documented that aging increases PFJ stress waveform, particularly late in stance, during stair ascent, but not descent; whereas navigating the uneven surface increases the PFJ stress waveform for both stair ascent and descent. These findings can be used to by clinicians to reduce risk of PFJ pain and injury for older adults and individuals frequently negotiating uneven stairs, alike. Specifically, these outcomes can be used to target specific knee biomechanics across all of stance, rather than weight acceptance exclusively, that may enhance injury risk for populations vulnerable to knee pain. Successful implementation of the knowledge provided herein by clinicians may result in a substantial reduction in the number of older adults that suffer PFJ pain during activities of daily living and decrease the $635 billion annually spent treating musculoskeletal pain.[2]

Limitations

This study may be limited by the PFJ stress calculation. Although accounting for sagittal knee kinematics and kinetics, the current PFJ stress model may be underestimating the muscle force applied across the knee joint, and subsequently stress, as the model does not account for hamstring muscle force or include rotational forces that may influence sagittal knee biomechanics. Further, the present model calculates PFJ contact area as a function of knee flexion angle, rather than participant-specific radiographic imaging (e.g., MRI). Previously, however, the estimation of PFJ reaction force and PFJ contact area have been compared to cadaveric and radiographically-derived measures with a reported r^2 of 0.99 according to van Eijden et al and Connolly et al.[126-129] Although we are confident using the Eijden et al. and Connolly et al. regression equations in our PFJ stress calculation accurately represents PFJ loading, using participant-specific models, accounting for individual variation in patellar shape (i.e., Type I, II, and III) and tracking, may provide a more accurate representation of PFJ contact area and improve our calculation, particularly at knee flexion angles that alter the patellofemoral articulation and subsequent PFJ contact area.[129]

The chosen challenging surfaces may be a limitation. Although intended to imitate shoes on ice, the coefficient of friction on the slick surface is still almost double that of a real-world setting (0.19 vs 0.10) and may not elicit similar gait adaptations as real ice. Further, the staggered wooden blocks of the uneven surface may have been predictable and did not accurately imitate the randomness of real-world uneven terrain. Although we observed gait differences for participants on both surfaces, a real-world setting may further exacerbate these differences and provide an inimitable effect on PFJ loading.

The chosen subject population may be a limitation. For the current study, we recruited older adults that have experienced at least one accidental fall in the last 12 months. However, older adult fallers may or may not exhibit differences in knee biomechanics than non-fallers. Regardless, understanding the biomechanical alterations adopted by older adult fallers on challenging surfaces provides valuable insights to the neuromuscular strategies adopted to safely perform activities of daily living, particularly under challenging conditions.

Finally, the self-selected walking speed may be a limitation. Although walk speed is correlated with vertical ground reaction forces and subsequent joint kinetics that may impact PFJ stress[107], the fact there was no significant difference in walk speed between the tested cohorts, we are confident this did not limit our findings.

Future Work

Aging and surfaces altered PFJ stress during stair negotiation tasks. As such, future research is warranted to determine if other activities of daily living (i.e., walking or pivoting) also increases PFJ stress and pain risk. Further, incorporating a wider range of challenging surfaces may enhance understanding of knee biomechanical alterations exhibited during real-world performance of activities of daily living.

Considering current participants exhibited waveform, but not magnitude differences in PFJ stress, expanding the waveform analysis to each of the variables associated to PFJ stress is warranted. Such knowledge may identify specific model variables (i.e., knee flexion angle, moment, PFJ reaction force, and PFJ contact area) that contribute to changes observed in PFJ stress and knee pain risk.

In addition, incorporating participant-specific models in to the PFJ stress calculation is warranted and may improve the estimation oof PFJ contact area, providing a more accurate representation of the PFJ stress waveform throughout the full knee range of motion of stance.

Lastly, replicating the current work using participants with non-fallers and individuals with PFJ pain or diagnosed PFJ osteoarthritis is also warranted. This might provide additional insight into how ailing populations adapt to stairs and challenging surfaces, and the explicit neuromechanical strategies to target to reduce pain and injury risk.

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APPENDIX A

Participant Strength

Maximal isometric hip and knee flexion and extension, and ankle dorsi- and plantar-flexion strength were recorded on an isokinetic dynamometer (HUMAC NORM, CSMI, Stoughton, MA, USA) for all participants.

For analysis, peak normalized hip, knee, and ankle flexor and extensor strength were submitted to an independent t-test to determine age differences.

Results

Older adults exhibited significantly lower peak knee and ankle muscle strength for both flexion and extension (all: $p < 0.009$). No significant difference was observed for peak hip flexion and extension ($p > 0.05$) (Table A.1).

Table A.1 Mean (SD) for peak lower limb isometric strength (Nm/kg) between young and older adults.

	Hip		Knee		Ankle	
	Flexion	**Extension**	**Flexion**	**Extension**	**Dorsiflexion**	**Plantarflexion**
Young Adults	1.55 (0.53)	1.84 (0.61)	1.50 (0.55)	2.71 (0.91)	0.51 (0.13)	1.03 (0.35)
Older Adults	1.21 (0.32)	1.45 (0.39)	1.00 (0.23)	1.78 (0.46)	0.35(0.12)	0.56 (0.22)
p-value	0.070	0.071	0.009	0.005	0.005	0.001

APPENDIX B

Waveform Analysis

Statistical parametric mapping (SPM) was used to analyze specific temporal differences in the PFJ stress waveforms. Movement analysis frequently compares discrete parameters between groups, such as magnitude or values at particular instants of task cycles. However, using predetermined parameters to test hypotheses leaves the results susceptible to biases, or more specifically regional focus bias and inter-component covariation bias.[148,149] By reducing the analysis to scalar parameters (0-dimensional), the entire time-series in not accounted for and, thus, differences at other instances of the task may be ignored (regional focus bias).[148,149] Also, components that describe joint movement may not be independent, so analyzing scalar components separately ignores signal covariance (inter-component covariation bas).[148] Thus, expanding the analysis over the entire time series of a given task could provide a better clinical understanding of the strategies adopted by different populations (i.e., older vs young adults) to achieve the same functional task. SPM uses Random Field Theory to determine statistical inference over 1-dimensional continuous vector trajectories and its application has been previously validated.[131,132] This technique will provide an understanding of specific temporal regions where significant differences between groups may occur, allowing for deeper analysis of PFJ stress waveforms over the entire task interval. Specifically, SPM will compare the instantaneous magnitudes at each point across the entire stance phase waveform between each group to identify specific "supra-threshold clusters" or regions where the groups and/or conditions are different from one another. This analysis, for instance, will identify regions the magnitude of the PFJ stress waveform (i.e., 50-80% of stance) statistically differs between young and older adults, or between surfaces, and provide additional information beyond traditional analysis of discrete point (i.e., peak) measures of PFJ stress.